(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
5 July 2001 (05.07.2001)

PCT

(10) International Publication Number
WO 01/47596 A2

(51) International Patent Classification[7]: A61N

(21) International Application Number: PCT/US00/41915

(22) International Filing Date: 6 November 2000 (06.11.2000)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
09/434,958 5 November 1999 (05.11.1999) US

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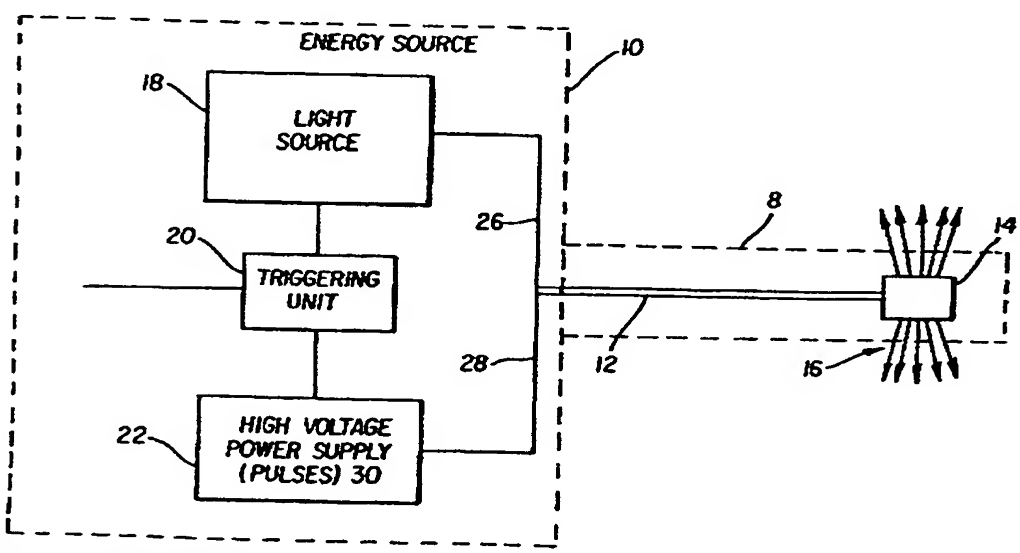
(81) Designated States (*national*): AE, AG, AL, AM, AT, AU, AZ, BA, BB, BG, BR, BY, BZ, CA, CH, CN, CR, CU, CZ, DE, DK, DM, DZ, EE, ES, FI, GB, GD, GE, GH, GM, HR, HU, ID, IL, IN, IS, JP, KE, KG, KP, KR, KZ, LC, LK, LR, LS, LT, LU, LV, MA, MD, MG, MK, MN, MW, MX, MZ, NO, NZ, PL, PT, RO, RU, SD, SE, SG, SI, SK, SL, TJ, TM, TR, TT, TZ, UA, UG, UZ, VN, YU, ZA, ZW.

(84) Designated States (*regional*): ARIPO patent (GH, GM, KE, LS, MW, MZ, SD, SL, SZ, TZ, UG, ZW), Eurasian patent (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European patent (AT, BE, CH, CY, DE, DK, ES, FI, FR, GB, GR, IE, IT, LU, MC, NL, PT, SE, TR), OAPI patent (BF, BJ, CF, CG, CI, CM, GA, GN, GW, ML, MR, NE, SN, TD, TG).

Published:
— *Without international search report and to be republished upon receipt of that report.*

For two-letter codes and other abbreviations, refer to the "Guidance Notes on Codes and Abbreviations" appearing at the beginning of each regular issue of the PCT Gazette.

(54) Title: APPARATUS AND METHOD FOR IN-SITU RADIATION TREATMENT



(57) Abstract: A miniaturized energy transducer that emits x-ray radiation is coupled to a catheter shaft to permit x-ray radiation treatment within a human body. The catheter shaft incorporates optical fibers and/or electrical conductors to supply optical and/or electrical energy to the energy transducer. Use of the miniaturized energy transducer in combination with the catheter shaft eliminates most of the problems related to the methods based on the use of radioactive transducers and offers a method for efficient and controllable radiation treatment. Specifically, a catheter shaft and energy transducer are designed to provide access to very narrow blood vessels for x-ray radiation treatment.

WO 01/47596 A2

APPARATUS AND METHOD FOR IN-SITU RADIATION TREATMENT

Field of the Invention

The present invention relates to an apparatus and method for in-situ radiation treatment in humans. More specifically, the invention provides a catheter device for insertion into human body cavities or blood vessels that includes a miniature x-ray source. Electrical and/or optical energy signals are supplied to the miniature x-ray source via a catheter shaft that includes electrical conductors and/or optical fibers.

Background of the Invention

Restenosis is a heart condition that afflicts 35%-50% of all people who undergo a balloon angioplasty to improve blood flow in narrowed sclerotic arteries. The condition consists of a significant re-closing of the treated artery segment hours to days after the procedure. As a result, the arterial lumen size is decreased and the blood flow downstream from the lesion site is impaired. Consequently, patients afflicted with restenosis must undergo an additional balloon angioplasty, and in some cases a coronary bypass surgery must be performed. Aside from pain and suffering inflicted on patients, recurrent stenosis is also a serious economic burden on society, with estimated expenses related to restenosis amounting to an estimated 0.8 - 3.0 billion dollars per year in the United States economy alone.

Attempts to treat restenosis have been concentrated in both the pharmacological and medical device areas. While pharmacological solutions have not been proven effective, some progress has been made with medical devices. Stents can be permanently inserted into an occluded artery to hold it open. Stents have been shown to be a sufficient remedy for two of the three mechanisms which cause recurrent stenosis, namely, elastic recoil of the artery and negative remodeling of the arterial structure. The third mechanism, neointimal growth, is a proliferation of smooth muscle cells from the lesion into the lumen and is not prevented by stents.

The greatest promise in avoiding recurrent stenosis appears to be ionizing radiation. Research has shown that gamma radiation delivered at the location of blood vessel treatment is effective in stopping both animal and human restenosis. In addition, locally delivered beta radiation has been shown to be effective in animal models,

wherein the most effective yet non-hazardous dosing rate is somewhere between seven and forty Gray (mJoule/gram).

In view of the above, various methods have been proposed to provide ionizing radiation treatment. For example, radiation catheters, based on the use of radioactive sources such as beta - emitting ^{32}P, $^{90}Sr/^{90}Y$, $^{188}W/^{188}Re$, beta+ emitting ^{48}V or gamma emitting ^{192}Ir, are at various stages of development and implementation. The radioactive sources, in a variety of configurations, are introduced via special catheters into the blood vessel and the radioactive source is placed at the treatment position for a predetermined time period for obtaining the proper irradiation dose. Another proposed method utilizes the implantation of a radioactive stent based on the above radioactive isotopes.

The gamma and beta radioactive sources used by the present radiation catheters and radioactive stents, however, have several drawbacks including the limited ability to provide selective control of time dosage or radiation intensity, and the logistical, regulatory, and procedural difficulties involved in dealing with radioactive materials. In addition, these devices expose healthy organs to dangerous radiation during the introduction of the radiation source and require hospital personnel to handle radioactive materials. Accordingly, use of these devices involves hazards to both the individual handling the radioactive materials and to the environment, and invokes nuclear regulatory requirements.

An alternative approach to providing ionizing radiation treatment is through the use of x-ray sources. Conventional x-ray radiation for radiotherapy is produced by high energy electrons generated and accelerated in a vacuum to impact on a metal target. The efficiency of x-ray generation is dependent on the electron beam current and on the acceleration voltage. Another method for the production of x-rays is by direct conversion of light into x-ray radiation. It is known that the interaction of light with a target can produce highly energetic x-rays when the power densities achieved are in the range of 10^{16}-10^{17} watt/cm^2. With the development of femtosecond laser, such power densities are achievable with moderate size lasers (See C. Tillman et al, NIMS in Phys. Res. A394 (1997), 387-396 and US Patent No. 5,606,588 issued to Umstadter et al., the contents of each of which are incorporated herein by reference). A 100

femtosecond pulse of 1mJ laser pulse focused down to a 3 micron spot, for example, will reach this power density level.

Medical applications of the direct conversion method of x-ray generation are currently in the development stage. The direct conversion method, for example, has been considered for medical imaging (See, Herrlin K et al. Radiology (USA), vol. 189, no. 1, pp. 65-8, Oct. 1993). Another medical application of femtosecond lasers is in improved non-thermal ablation of neural or eye tissue for surgical purposes (See, F.H. Loesel et al. Appl.Phys.B 66,121-128 (1998)). The development of compact table top models of femtosecond lasers makes the radiotherapy application of laser generated x-rays an attractive alternative.

In view of the above, it is an object of the present invention to provide a non-radioactive source that can be inserted into the body by a catheter device to emit a controlled dose of radiation with equivalent or better biological effectiveness than intra vascular gamma and beta sources, without using radioactive source materials. It is a further object of the present invention to provide a catheter device with a radiation source that can be moved at a precisely controlled rate, controlled to vary the amount of radiation emitted at a desired location within the blood vessel, and controlled to vary the penetration depth of the radiation.

Summary of the Invention

The present invention relates to miniaturized energy transducers that are coupled to catheters to permit x-ray radiation treatment within a human body. The present invention eliminates most of the problems related to the methods based on the use of radioactive sources and offers a method for efficient and controllable radiation treatment. Specifically, a catheter and energy transducer design is provided that has a diameter of 1.7 mm or less, and most preferably about 1 mm, thereby allowing access to very narrow blood vessels for x-ray radiation treatment.

More specifically, in one preferred embodiment, an x-ray transducer is provided that includes an electrically insulating tube, a cathode provided at a first end of the electrically insulating tube, an anode provided at a second end of the electrically insulating tube, and an outer conductive layer, located on an outer surface of the

insulating tube and electrically coupled to the anode, that extends from the second end of the electrically insulating tube to the first end of the electrically insulating tube.

The cathode preferably includes a conducting cathode shell that forms a cavity. In one example, a light pulse is applied to the conducting cathode shell in order to heat an outer surface of the conducting cathode shell to cause thermionic emission of electrons from the outer surface. In another example, the conducting cathode shell includes an electron escape nozzle, and a plasma is generated in the cavity either by applying a light pulse to an inner surface of the conducting cathode shell or by providing a spark gap in the cavity of the conducting cathode shell. A mechanism is provided for applying a voltage pulse to the anode at least one of during a period and immediately after a period in which electrons are emitted by the cathode.

In a further embodiment, an x-ray transducer includes an insulating shell that forms a cavity, a conducting anode located at a first end of the cavity of the insulating shell, and an emission element located at a second end of the cavity opposite the conducting anode. The emission element is either a photo-emission electron source or a thermionic emission surface.

In another embodiment, an x-ray transducer includes conducting anode that forms a cavity, a conducting cathode located within the cavity, and a radial insulator provided between the conducting anode and conducting cathode.

A variety of catheter designs are provided, including one having an optical fiber central core, a conducting inner layer formed on the central core, an insulator surrounding the conducting inner layer; and a conducting outer layer formed on the insulator. The central core has a diameter in a range from about 50 to about 500 microns. The conducting inner layers have a thickness of between about 10 to about 100 microns. The insulator comprises at least one of plastic, Teflon™ and a polymer.

In a further embodiment, a catheter includes a conductive core, a bundle of optical fibers embedded in a polymer insulation around the core, and an outer ring of conductive wires embedded in the polymer insulation. The core comprises a metallic conductor having a diameter of between about 50 and about 500 microns. The diameter of each of the optical fibers range from about 10 microns to about 200 microns. The

conductive wires are made from a metallic material having a diameter in a range from approximately 10 to 100 microns.

In a still further embodiment, a catheter is provided that includes a conducting core, a bundle of optical fibers embedded in a polymer insulation around the core, an outer ring of insulation around the optical fibers, and a conducting outer layer around the outer ring of insulation. The core comprises a metallic conductor having a diameter in a range between approximately 50 and 500 microns. Each of the optical fibers have a diameter in a range from approximately 10 microns to 200 microns. The outer ring of insulation has a thickness in a range from approximately 100 to 500 microns. The conducting outer layer comprises a metal having a thickness in a range from approximately 10 to 100 microns.

A primary feature of the invention is the use of energy source that provides voltage pulses that are applied to the anode, which increases the electrical breakdown threshold of the device.

Other advantages and features of the invention will become apparent from the following detailed description of the preferred embodiments and the accompanying drawings.

Brief Description of the Drawings

The invention will now be described with reference to certain preferred embodiments thereof and the accompanying drawings, wherein:

Fig. 1 is a functional block diagram of a radiotherapy system in accordance with the present invention;

Fig. 2 is a cross-sectional view of a linear type energy transducer in accordance with the invention;

Fig. 3 is a cross-sectional view of an energy transducer that includes a laser induced plasma cathode;

Fig. 4 is a cross-sectional view of an energy transducer that includes a laser induced thermionic cathode;

Fig. 5 is a cross-sectional view of a liner reverse energy transducer in accordance with the invention;

Fig. 6 is a cross-sectional view of an energy transducer that includes a spark induced plasma cathode;

Fig. 7 is a cross-sectional view of a radial type energy transducer in accordance with the invention;

Fig. 8 is a cross-sectional view of a preferred catheter shaft including a coaxial catheter with a central optical fiber core;

Fig. 9 is a cross-sectional view of a preferred catheter shaft including external conducting wires;

Fig. 10 is a cross-sectional view of a preferred catheter shaft including a bundle with an external conducting coating;

Fig. 11 is a cross-sectional view of an x-ray emitting catheter device including a catheter shaft that incorporates memory alloy conductors;

Fig. 12 illustrates the device of Fig. 11 with a distal nitinol tube;

Fig. 13 illustrates the combination of a catheter device in accordance with the invention with an angioplasty balloon; and

Fig. 14 illustrates a further embodiment of the invention that employs flexible centering wires.

Detailed Description of the Preferred Embodiments

Fig. 1 illustrates a radiotherapy treatment system in accordance with the present invention. The radiotherapy treatment system includes an energy source 10 coupled to a catheter device 8. The catheter device 8 includes a catheter shaft 12 and a miniature energy transducer 14 located at a proximal end of the catheter shaft 12. The catheter shaft 12 delivers energy from the energy source 10 to the miniature energy transducer 14, which preferably converts electrical and/or optical energy received form the energy source 10 into x-ray radiation and distributes the x-ray radiation (illustrated by arrows 16) in a predetermined distribution pattern. The energy source 10 is preferably located external to the patient, while the catheter shaft 12 is manipulated to place the energy transducer 14 in an area to be treated within the body of a patient.

In general, the energy source 10 is adapted to provide electrical and/or optical energy through the catheter shaft 12 that is correspondingly configured to deliver the

energy to the energy transducer 14. Accordingly, the energy source 10 is provided with a light source 18, for example a laser, and a power supply 22, such as voltage pulse generator, respectively connected through optical and electrical conductors 26, 28 to the catheter shaft 12. A triggering control unit 20 directs the energy source 10 to deliver optical and electrical energy through the catheter shaft 12 to the energy transducer 14 as required by the operator. The duration and amplitude of the pulses applied by the light source 18 and the power supply 22, for example, can be varied to control the distribution of the x-ray radiation produced by the energy transducer 14.

The energy transducer 14 is preferably surrounded by x-ray transmissive insulation (not shown) that can be presented in direct contact with the human body. The transmissive insulation may be a material coated on an outer surface of the energy transducer 14. Alternatively, the transmissive insulation may take the form of a capsule that encapsulates the energy transducer 14. In any case, the energy transducer 14 is preferably a relatively low-cost, replaceable and disposable unit which avoids the necessity of complex sterilization processes required for instruments that are intended for multiple use.

Referring now to Fig. 2, a first preferred embodiment of the energy transducer 14 includes a conducting cathode 30 and a conducting anode 32 respectively located at a proximal end and a distal end of an electrically insulating tube 34. The sealed tube 34 is preferably 3-9 mm in length and 0.9-1.5 mm in diameter, and holds a vacuum within a hollow chamber 38 that typically varies from 10^{-3} to 10^{-6} Torr depending on the type of electron generation method employed. The proximal end of the energy transducer 14 is coupled to the catheter shaft 12, which includes conductors that provide electrical and/or optical energy to the conducting cathode 30 as will be described in greater detail. In order for an electrical connection to be provided to the conducting anode 32 at the distal end of the energy transducer 14, an outer conductive layer 36 is provided on the outer surface of the insulating tube 34, which connects with a conductor provided in the catheter shaft 12.

In operation, electrons are generated at the conducting cathode 30 upon the application of an energy signal received from the energy source 10 while the conducting anode 32 is held at ground potential. Electron generation at the cathode generally takes about 10 nsec, during which time or immediately thereafter, a negative high-voltage pulse is introduced to the conducting cathode 30. The magnitude of the pulse may vary between

-15 kV and -50 kV and have a duration between 20 nsec to 200 nsec. The electrons generated at the conducting cathode 30 are accelerated across the tube 34 by the application of the voltage pulse to the conducting cathode 30, until they are stopped by the conducting anode 32 which results in the generation of the required x-rays. The conducting anode 32 and the conducting cathode 30 are held at ground potential once the pulse applied to the conducting cathode 32 terminates. The process is then repeated until the desired results are obtained. The pulse rate for the pulses applied to the conducting cathode 30 may vary from 1 Hz to 1000 Hz.

The conducting cathode 30 may generate the required electrons utilizing a variety of different methods. Fig. 3, for example, illustrates a conducting cathode 30 that includes a conducting cathode shell 40 that forms a cavity 42. The conducting cathode shell 40 is provided with an electron escape nozzle 44 having a diameter in the range of about 10 to 200 microns. In this embodiment, laser light is introduced into a proximal end of the conducting cathode shell 40 via an optical fiber 46 provided within the catheter shaft 12. The conducting cathode shell 40 is also electrically coupled to an electrical conductor 47 provided around the optical fiber 46 within the catheter shaft 12. In this illustration, for simplicity, the conductor in the catheter shaft 12 coupled to the conducting anode 32 is not shown. The laser light may optionally be focused by a lens 48 provided at the end of the optical fiber 46. The lens 48 may constitute a separate element or the end of the optical fiber 46 can be shaped in the form of a lens. The laser light strikes the distal end of the cavity 42 inside the conducting cathode shell 40. When high laser intensities, greater than $100 MW/cm^{2}$, are introduced into the conducting cathode shell 40, a plasma containing electrons, ions and neutral atoms is formed, for example, by the ablation of the material that composes the conducting cathode shell 40 at the point of impact. In a preferred embodiment, a laser light pulse is applied that has a duration of 1-20 nsec and an intensity of 10^{9} - 10^{11} W/cm^{2}.

The electrons possess a thermal velocity that is approximately 50 times greater than that of the ions. As a result, a significant percentage of the electrons escape through the electron escape nozzle 44 before a considerable number of ions escape. The electrons that escape are replenished from the cavity walls of the conducting cathode shell 40 due to thermionic emission and/or photo emission induced by the high temperature of the plasma. The small diameter of the electron escape nozzle 44 effectively renders the cavity 42 a

Faraday cage, thereby isolating electrons in the chamber 38 from fields formed by charges inside the cavity 42 and vice versa.

The electron escape nozzle 44, however, can also be eliminated as illustrated in the embodiment shown in Fig. 4. In this embodiment, a conducting cathode shell 50 is also provided that forms a cavity 52. Laser energy is introduced via an optical fiber 54 provided in the catheter shaft 12, and the conducting cathode shell 50 is coupled to a conductor 55 provided on the optical fiber 54. Again a lens 56 may optionally be used to focus the laser energy to heat the distal outer surface 58 of the conducting cathode shell 50. The temperature of the outer surface 58 is raised to a level comparable with the cathode material's work function and, as a result, electrons are emitted from the distal end of the conducting cathode shell 50. The laser light pulse is applied in the same manner discussed above.

A second preferred embodiment of an energy transducer 14 in accordance with the invention is illustrated in Fig. 5. As in the previously described embodiment, the catheter shaft 12 is provided with an internal optical fiber 60 having a lens 62 located at its distal end. An x-ray transmissive insulating tube or shell 64 is provided around the end of the optical fiber 60 to form a cavity 66 containing a conducting anode 68 and an emission element 70 which serves as the cathode. The conducting anode 68 is coupled to a conductor 61 provided around the optical fiber 60. The emission element 70, for example, is a photo-emission cathode or thermionic emission surface, and is grounded by a metallic layer that contacts with a conductor in the catheter shaft 12. The insulating shell 64 is sealed and holds a vacuum that typically varies from 10^{-6} Torr, when a photo cathode is used for the emission element 70, to 10^{-4} Torr, when a thermionic emission surface is used for the emission element 70.

In this embodiment, electrons are emitted by the photoelectric effect when a photocathode is used as the emission element 70, and electrons are emitted by the thermionic emission effect when a thermionic emission surface is used as the emission element 70. Laser light is focused on the emission element 70 by the lens 62 (although the lens 62 can be omitted if desired), wherein the electron generation process preferably takes approximately 10 nsec. The same pulse duration and amplitude are utilized as with the previously described embodiments. Immediately thereafter, or even during the process of electron generation, a positive high-voltage pulse is introduced to the conducting anode 60.

The magnitude of the pulse may vary between +15kV and +50kV. The electrons created at the emission element 70 are accelerated across the cavity 66 by the pulse and are stopped at the conducting anode 62 thereby causing the required x-rays to be emitted.

If the emission element 70 is a photo cathode, it is preferably made of a highly efficient photo emissive material, with an optimal wave length efficiency response, conventional materials of the type suitable for this purpose include, but are not limited to, metals, such as Au, Mg, Cu, semiconductors, like gallium arsenic, and compounds, such as Cs3 Sb, Cs2 Te, and AgOCs. It will be appreciated by those skilled in the art that such photo emissive materials vary in efficiency and in operating conditions parameters, such as vacuum requirements and the light wave length of the light source, which are well known.

If the emission element 70 is a thermionic emission surface, it is preferably made of tungsten, tantalum or other refractory metals having high melting and sublimation points. Electron currents greater that 500 Amps/cm^2 have been measured experimentally in tantulum and tungsten. Additionally, the ratio between the work function and the melting temperature if low for these materials, which renders them good thermionic emitters.

It is also possible to provide a miniature energy transducer that emits x-rays without the use of a light source as required in the previously described embodiments. Fig. 6, for example, illustrates a spark induced plasma cathode 80 that can be utilized in the linear type miniature energy transducer 14 illustrated in Fig. 2. in accordance with a third preferred embodiment of the invention. In this embodiment, the catheter shaft 12 has internal electrical conductors 72 and 74 that are separated by an insulator 76, thereby creating a spark gap 78 at a distal end of the catheter shaft 12. A conducting cathode 80 is provided that, like the embodiment of Fig. 3, includes an electron escape nozzle 82 having a diameter of approximately 10 to 200 microns. In this embodiment, high voltage pulse, of a duration between 10-200 nsec and a magnitude of 5-50 kV, is applied across the spark gap 78 via the internal electrical conductors 72, 74 of the catheter shaft 12. The electrical breakdown creates hot plasma, consisting of electrons, ions and neutral atoms, that is formed within a cavity 62 formed by the cathode 80. As in the previous embodiment, the electrons possess a thermal velocity that is 50 times greater than that of the ions. As a result, a significant percentage of the electrons escape through the electron

escape nozzle 82 before a considerable number of ions escape. The electrons that escape are replenished from the cavity 62 walls by thermionic emission and photo-emission induced by the plasma.

A fourth embodiment of a miniaturized energy transducer in accordance with the invention, which will be referred to as a radial embodiment, is illustrated in Fig. 7. The energy transducer includes a conducting cathode 84, a conducting anode 86 that also serves as an outer shell to form a cavity 88, and radial insulation 90 located between the conducting anode 86 and conducting cathode 84. The conducting cathode 84 can be either of a spark-induced type, laser induced plasma type or laser induced thermionic emission type as described above with respect to the other embodiments. The cavity 88 is sealed and holds a vacuum as with the previously described embodiments.

In operation, as with previous embodiments, electrons are generated at the conducting cathode 84. As electron generation is taking place or immediately thereafter, a negative high-voltage pulse is introduced to the conducting cathode 86. As described above, the magnitude of the pulse may vary between -15 kV and -50 kV. The electrons are accelerated radially from the tip of the conducting cathode 84 by the high voltage, and are stopped at the conducting anode 86 thereby causing the emission of the required x-rays. The internal portion of the insulation 90 does not touch the conducting cathode 84 and conducting anode 86. This increases the surface area of the insulation 90 between points of high voltage and ground, thus reducing the probability of electrical breakdown along the surface of the insulation 90. The advantages of this embodiment over the aforementioned ones is that there is no risk that electrons will short the circuit by breaking through the insulating tube 34 of the other embodiments into the outer conducting layer 36 held at ground.

It should be noted that in the preferred embodiments, the portion of the energy transducer 14 that is in direct contact with the patients body or blood is preferably kept at ground potential, while the pulse applied to accelerate the electrons is applied to the element that is insulated from the patient. For example, in the linear embodiment illustrated in Fig. 2, a negative voltage pulse is applied to the conducting cathode 30 while the conducting anode is grounded. However, in the reverse linear embodiment illustrated in Fig. 5, a positive voltage pulse is applied to the conducting anode 68 while the emission element 70 is grounded. In the radial embodiment illustrated in Fig. 7, the conducting

cathode 84 again receives a negative voltage pulse while the conducting anode 86 is grounded.

As can be seen by the illustrated examples, the miniature energy transducer 14 may consist of a variety of different types, each of which can utilize a pulsed energy source to accelerate the electrons. Further, the catheter shaft 12 not only serves the purpose of inserting and extracting the miniature energy transducer 14 into an out of the body, but also is required to supply the high voltage pulses and/or optical pulses required from the high voltage power supply 22 and the light source 18. The catheter shaft 12 preferably combines the electrical and optical transmission lines into one structure, thus economizing the catheter diameter that must be preferably 1.7 mm or less, and most preferably about 1.0 mm, for coronary applications, and thereby allowing the catheter shaft 12 to follow the contours of a blood vessel or any other body cavity. Further, the catheter shaft 12 must be flexible enough to negotiate the sharp bends in the vascular system while maintaining its electrical and/or light conducting properties. Accordingly, in addition to providing several different alternatives for the miniature energy transducer 14, it is also preferable to provide several different alternatives for the structure of the catheter shaft 12.

Fig. 8 illustrates a first embodiment of a preferred catheter shaft design incorporating a coaxial catheter bundle. In this embodiment, the catheter shaft 12 has a central core made out of either a glass or plastic optical fiber 92. The diameter of the core optical fiber 92 may range from 50 to 500 microns. The core optical fiber 92 is coated with a layer of conducting metal to form a conducting inner layer 94 preferably 10 to 100 microns thick. An insulator 96 surrounds the conducting inner layer 94, and is preferably plastic, Teflon™, or another polymer. A conducting outer layer 98 surrounds the insulator 96. In a preferred embodiment, the optical fiber 92 is terminated in a spherical fashion at its distal end.

The catheter shaft 12 illustrated in Fig. 8 can be utilized, for example, in conjunction with the miniature energy transducer 14 illustrated in Fig. 2. In such a case, the conducting inner layer 94 is preferably connected to the conducting cathode 30, the conducting outer layer 98 is connected to the conducting anode 32, and the core optical fiber 92 provides the optical signals required by the cathode embodiments illustrated in Figs. 3 and 4. The coaxial design serves as a nearly perfect wave-guide, enabling the transmission of high voltage pulses without distortion.

Fig. 9 illustrates a second type of preferred catheter shaft with conducting wires. A conducting core 100 of the catheter shaft 12 is made out of a metallic conductor with a diameter between 50 and 500 microns. A bundle of optical fibers 102 are embedded in a polymer insulator 104 around the conducting core 100. The diameter of each of the optical fibers 102 can range from 10 microns to a few hundred microns, and the number of optical fibers 102 in the bundle can range from a few to a few hundred fibers. External to the optical fibers 102, an outer ring of conducting outer wires 106 are also embedded into the insulator 104. The conducting outer wires 106 are made from a metal material and their diameter may range from approximately 20 to 50 microns. The conducting outer wires 106 are preferably at ground potential while the conducting core 100 is used to carry the high voltage pulses. The ring of conducting outer wires 106 and the conducting core 100 serve as the conductors of a coaxial cable. This design provides sufficient flexibility to negotiate the curves of the vascular system.

Fig. 10 illustrates a further preferred catheter shaft embodiment. In this embodiment, as with the previous embodiment just described, a conducting core 108 is made out of a metallic conductor with a diameter between approximately 50 and 500 microns and a bundle of optical fibers 110 are embedded in polymer insulator 112. The diameter of each of the optical fibers 110 can range from about 10 microns to a few hundred microns, and the number of fibers can again vary. External to the optical fibers 110, an outer ring of insulation 114 is provided that does not contain embedded optical fibers. The thickness of the outer ring of insulation 114 preferably ranges from about 100 to 500 microns. A conducting outer layer 116, made out of a metal having a thickness range from about 10 to 100 microns, is provided around the outer ring on insulation 114.

The present invention has applications in a wide range of minimally invasive brachytherapy procedures such as intra vascular irradiation treatment for the prevention of restenosis following angioplasty, treatment of organs and implantation in organs, tumor irradiation and other irradiation applications in the human body. The variety of available catheter and energy transducer designs permits different types of configurations to be utilized for different types of procedures, thereby insuring that clinical results are maximized. Accordingly, the present invention further provides a method for the generation of x-rays in situ in the human body for minimally invasive radiation treatment of various conditions, and particularly for treatment for the prevention of restenosis in the

human body. The method includes locating the energy transducer of the invention in a human body cavity at a point of treatment, such as a blood vessel, through the manipulation of the catheter shaft, activating the cathode of the energy transducer to generate electrons, either through the application of electrical energy or optical energy; applying a voltage pulse of a specified amplitude to either the cathode or anode of the energy transducer thereby causing the electrons to be drawn to the anode; grounding the cathode and anode after the application of the voltage pulse; and repeating the process for a specified dwell time. In such method, the amplitude of the voltage pulse can be controlled to vary the penetration depth of the x-rays, while the dosage amount can be varied by varying the dwell time, i.e., the time at which the energy transducer is located at the point of treatment.

As will be readily understood by those of ordinary skill in the art, it is desirable to maximize the flexibility of the catheter shaft 12. A further embodiment of a preferred catheter shaft and energy transducer is illustrated in Fig. 11. In this embodiment, a 3-5 nsec laser pulse is transmitted through an optical fiber 120 that serves as the core of the catheter shaft 12. The laser pulse exits the optical fiber 120 through a polished surface 122 or lens and ablates a tungsten, tantalum or other metallic cathode 124 to create a plasma in a cavity 126. Upon termination of the laser pulse, a -15 kV to -50 kV pulse with a 2-20 nsec duration is transmitted through a transmission line in the catheter shaft 12 that includes the inner conductor 130 and outer conductor 132 that are separated by a insulator 134 preferably made of polyiamide, plastic, Teflon™ or a polymer. The inner conductor 130 and outer conductor 132 are made of nitinol, a flexible shape memory alloy. The high voltage pulses accelerate the electrons to a distal tip cover 134 that serves as an anode, which is coupled to the outer conductor 132. The electrons attain an energy level around 15 KeV - 50 KeV before they are decelerated and stopped at a target 136, for example a tungsten coating 20 microns thick, located opposite the holes 128. A vacuum on the order of 10^{-4} Torr is provided in the cavity 126, which is necessary for the uninterrupted acceleration of the electrons by the high voltage.

A vacuum is applied to the cavity 126 through a hole 138 which is either sealed or connected to a distal nitinol tube 144, shown in Fig. 12, with a getter material provided at its end. The getter material and the volume of the distal nitinol tube 144 serve to preserve the quality of the vacuum. Because nitinol is flexible, the distal tube 144 does not hinder the catheter's ability to navigate the arterial system.

The inner and outer conductors 130 and 132 are made of nitinol only at the distal end of the catheter shaft 12. In all other regions, the conductors are made of aluminum or other metal. The transition between nitinol and aluminum preferably preferable occurs 16 mm before the distal tip. The inner conductor 130 extends into the energy transducer 14 to contact with the cathode 124. The outer conductor 132 contacts with the distal tip cover 134 constituting the anode.

As also shown in Fig. 11, a crease or gap 140 is provided in the insulator 134, and an inner layer 142 of the insulator 134 defined by the crease extends into the energy transducer 14. The crease 140 in the insulator 134 increases the over the surface distance from the most proximal of the holes 128 to about 10 mm, which assists in eliminating a flashover breakdown phenomenon. The nitinol inner and outer conductors 130 and 132 keeps the crease 140 from collapsing as the nitinol remembers its original straight shape.

The stiff energy transducer 14 has a length less than 3 mm in the illustrated embodiment with a diameter of only 1.5 mm. Accordingly, the use of the nitinol conductors and the crease allows the portion of the catheter device that is stiff to be significantly reduced, thereby allowing greater flexibility in the movement of the catheter device. However, once placed in position, the nitinol conductors straighten to their original configuration, thereby insuring proper orientation and placement of the energy transducer.

The invention provides many advantages over conventional treatment methods and devices. Experiments have shown that a 15 Gray doses of radiation can be delivered 2 mm from an arterial wall with a lesion length of 30 mm in less than 10 minutes for a non-optimized laser cathode, while similar doses can be applied in less than 40 minutes for a non-optimized electrical cathode. The use of pulsed voltage increases the breakdown voltage. Improved miniaturization is achieved with improved x-ray yield per energy input. The electron production at the cathode is independent from x-ray production at the anode. Electron production at the cathode is only weakly dependent on cathode size, accordingly it is believed the invention can be further scaled to produce a sub-millimeter diameter device.

The invention has been described with reference to certain preferred embodiments thereof. It will be understood, however, that modification and variations are possible within the scope of the appended claims. The shape and location of the anodes and

cathodes can be readily varied. In addition, while the light source is preferably a laser light source, a high intensity flash lamp may also be employed. Still further, it is desirable in some applications to control the radiation dose estimate, and it is possible to monitor in real time the radiation dose delivered by incorporating into the catheter device additional conducting leads for a dose sensor. A sensor suitable for this application is for example a RADFET manufactured by ESA-ESRIN real time integrating dosimeter, which can be mounted together with the energy transducer Another possibility is to incorporate a miniature MOSFET detector, produced for example by Thomsom and Nielson, Ottawa, Canada, or a Scanditronix p-type semiconductor detector, produced by Scanditronix, Uppsala, Sweden, into the energy transducer. Instead of a monitoring device, the the catheter shaft 12 and energy transducer 14 can also be combined with other medical procedure instrumentality. Fig. 13, for example, illustrates the incorporation of the catheter 12 and energy transducer 14 with a balloon 146 that is utilized for angioplasty in coronary blood vessels, wherein x-rays can be generated and the patient be irradiated simultaneously with an angioplasty procedure. Also, a motorized pull back mechanism can be employed to permit the operator to energize the system to commence emitting x-ray radiation and then control and vary the amount of time the tip dwells at each desired location within the blood vessel. Finally, flexible wires extending from the energy transducer may be utilized as a centering device. For example, as illustrated in Fig. 14, flexible wires 144 extend from the anode 146 of a linear type energy transducer that is located with a cathode 148 in an insulating tube 150. In this example, the flexible wires 144 can provide a dual purpose of providing a conductive connection to the catheter shaft 12 in place of the outer conducting layer 36 illustrated in Fig. 2. Three wires are preferably employed, each located 120 degrees apart.

WHAT IS CLAIMED IS:

1. An x-ray transducer comprising:

an electrically insulating member;

a cathode provided at a first end of the electrically insulating member;

an anode provided at a second end of the electrically insulating member; and

an outer conductor that extends from the second end of the electrically insulating member to the first end of the electrically insulating member.

2. An x-ray transducer as claimed in claim 1, wherein the cathode comprises a conducting cathode shell that forms a cavity.

3. An x-ray transducer as claimed in claim 2, wherein the conducting cathode shell includes an electron escape nozzle.

4. An x-ray transducer as claimed in claim 1, wherein an outer diameter of the x-ray transducer is 1.7 mm or less.

5. An x-ray transducer as claimed in claim 1, wherein the electrically insulating member is a cylindrical tube.

6. An x-ray transducer as claimed in claim 5, wherein the electrically insulating tube has a length from 3 to 7 mm and a diameter from 0.9 to 1.5 mm.

7. An x-ray transducer as claimed in claim 2, further comprising means for heating an outer surface of the conducting cathode shell to cause thermionic emission of electrons from the outer surface.

8. An x-ray transducer as claimed in claim 2, further comprising plasma generating means for generating a plasma within the cavity of the conducting cathode shell.

9. An x-ray transducer as claimed in claim 8, wherein the plasma generating means comprises light transducer means for applying a light pulse to an inner surface of the conducting cathode shell.

10. An x-ray transducer as claimed in claim 8, wherein the plasma generating means comprises means for providing a spark gap in the cavity of the conducting cathode shell.

11. An x-ray transducer as claimed in claim 1, further comprising means for applying a voltage pulse to the cathode at least one of during a period and immediately after a period in which electrons are emitted by the cathode.

12. An x-ray transducer comprising:

an insulating shell that forms a cavity;

a conducting anode located at a first end of the cavity of the insulating shell; and

an emission element located at a second end of the cavity opposite the conducting anode.

13. An x-ray transducer as claimed in claim 12, wherein the emission element comprises a photo-emission cathode.

14. An x-ray transducer as claimed in claim 12, wherein the emission element comprises a thermionic emission surface that acts as a cathode.

15. An x-ray transducer as claimed in claim 12, further comprising means for focusing a light pulse on the emission element.

16. An x-ray transducer as claimed in claim 12, wherein an outer diameter of the x-ray transducer is 1.7 mm or less.

17. An x-ray transducer as claimed in claim 12. further comprising means for applying a voltage pulse to the anode at least one of during a period and immediately after a period in which electrons are emitted by the emission element.

18. An x-ray transducer comprising:
a conducting anode that forms a cavity;
a conducting cathode located within the cavity; and
a radial insulator provided between the conducting anode and conducting cathode.

19. An x-ray transducer as claimed in claim 18, wherein the cathode comprises a conducting cathode shell that forms a cavity.

20. An x-ray transducer as claimed in claim 19, wherein the conducting cathode shell includes an electron escape nozzle.

21. An x-ray transducer as claimed in claim 18, wherein an outer diameter of the x-ray transducer is 1.7 mm or less.

22. An x-ray transducer as claimed in claim 19, further comprising means for heating an outer surface of the conducting cathode shell to cause thermionic emission of electrons from the outer surface.

23. An x-ray transducer as claimed in claim 20, further comprising plasma generating means for generating a plasma within the cavity of the conducting cathode shell.

24. An x-ray transducer as claimed in claim 23, wherein the plasma generating means comprises light transducer means for applying a light pulse to an inner surface of the conducting cathode shell.

25. An x-ray transducer as claimed in claim 24, wherein the plasma generating means comprises means for providing a spark gap in the cavity of the conducting cathode shell.

26. An x-ray transducer as claimed in claim 18, further comprising means for applying a voltage pulse to the cathode at least one of during a period and immediately after a period in which electrons are emitted by the cathode.

27. A catheter shaft comprising:
an optical fiber central core;
a conducting inner layer formed on the central core;
an insulator surrounding the conducting inner layer; and
a conducting outer layer formed on the insulator.

28. A catheter shaft as claimed in claim 27, wherein the central core has a diameter in a range from about 50 to about 500 microns.

29. A catheter shaft as claimed in claim 27, wherein the conducting inner layer has a thickness of between about 10 to about 100 microns.

30. A catheter shaft as claimed in claim 27, wherein the insulator comprises at least one of plastic, Teflon™ and a polymer.

31. A catheter shaft comprising:
a conductive core;
a bundle of optical fibers embedded in a polymer insulation around the core; and
an outer ring of conductive wires embedded in the polymer insulation.

32. A catheter shaft as claimed in claim 31, wherein the core comprises a metallic conductor having a diameter of between about 50 and about 500 microns.

33. A catheter shaft as claimed in claim 31, wherein the diameter of each of the optical fibers range from about 10 microns to about 200 microns.

32. A catheter shaft as claimed in claim 31, wherein the conductive wires are made from a metal material having a diameter in a range from approximately 10 to 100 microns.

33. A catheter shaft comprising:
a conducting core;
a bundle of optical fibers embedded in a polymer insulation around the core;
an outer ring of insulation around the optical fibers; and
a conducting outer layer around the outer ring of insulation.

34. A catheter shaft as claimed in claim 33, wherein the core comprises a metallic conductor having a diameter in a range between approximately 50 and 500 microns.

35. A catheter shaft as claimed in claim 33, wherein each of the optical fibers have a diameter in a range from approximately 10 microns to 200 microns.

36. A catheter shaft as claimed in claim 33, wherein the outer ring of insulation has a thickness in a range from approximately 100 to 500 microns.

37. A catheter as claimed in claim 33, wherein the conducting outer layer comprises a metal having a thickness in a range from approximately 10 to 100 microns.

38. An apparatus comprising:
a catheter shaft;
an energy transducer coupled to a distal end of the catheter shaft; and
means for providing pulsed energy to the energy transducer to generate x-rays.

39. An apparatus as claimed in claim 38, wherein the means for providing pulsed energy comprises a pulsed voltage transducer that is coupled to the energy transducer by an electrical conductor provided in the catheter shaft.

40. An apparatus as claimed in claim 38, wherein the means for providing pulsed energy comprises a pulsed light transducer that is coupled to the energy transducer by an optical fiber provided in the catheter shaft.

41. A method of generating x-rays within a human body, comprising:

locating an x-ray transducer at a treatment area by manipulation of a catheter shaft attached to the x-ray transducer; and

applying a series of energy pulses from an energy transducer to the x-ray transducer to cause the x-ray transducer to generate x-rays for a specified dwell time and at a specified voltage level.

42. A method of generating x-rays within the human body as claimed in claim 41, further comprising adjusting a dosage profile of the generated x-rays by varying at least one of the dwell time and the voltage level.

43. A catheter shaft as claimed in claim 27, wherein the inner conductor and outer conductor comprise a memory shape alloy for a specified distance from a distal end of the catheter shaft.

44. A catheter shaft as claimed in claim 43, wherein a crease is provided in the insulator that extends from the distal end of the catheter shaft.

45. An x-ray transducer as claimed in claim 1, wherein the outer conductor comprises a plurality of wires.

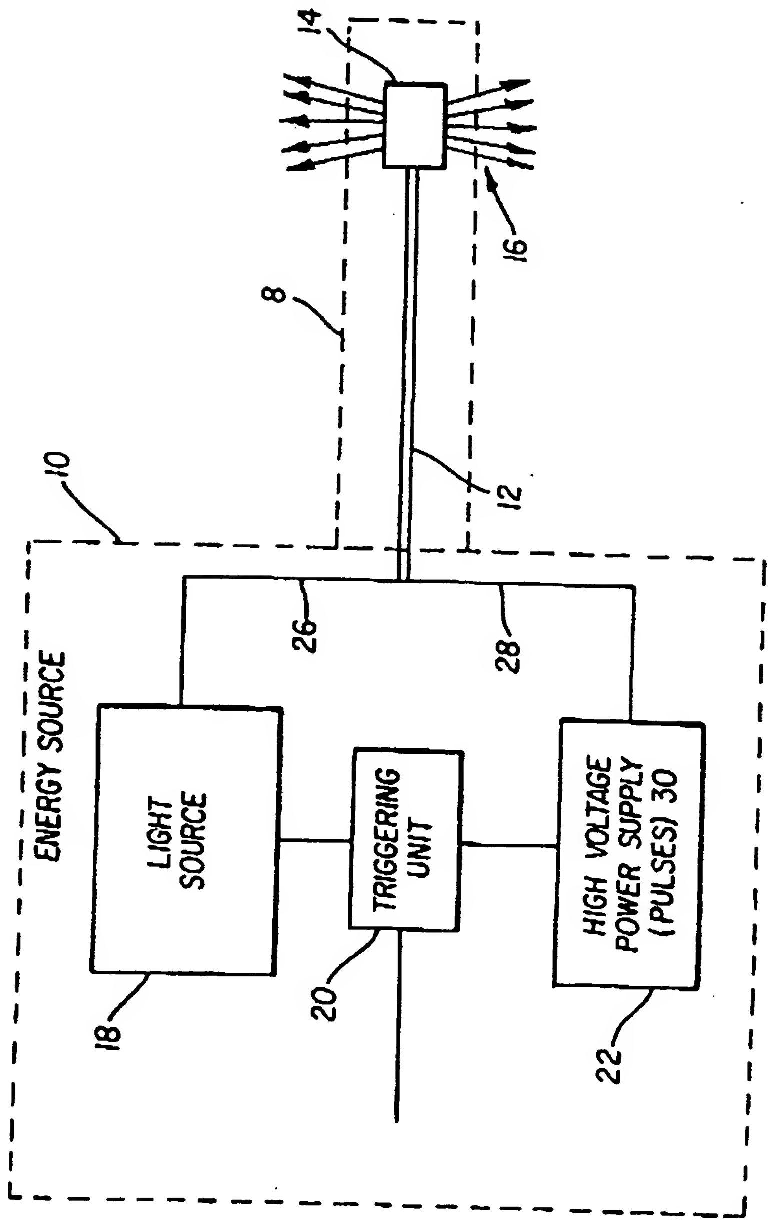

14
16
8
10
12
ENERGY SOURCE
26
28
LIGHT SOURCE
TRIGGERING UNIT
HIGH VOLTAGE POWER SUPPLY (PULSES) 30
18
20
22

FIG. 1

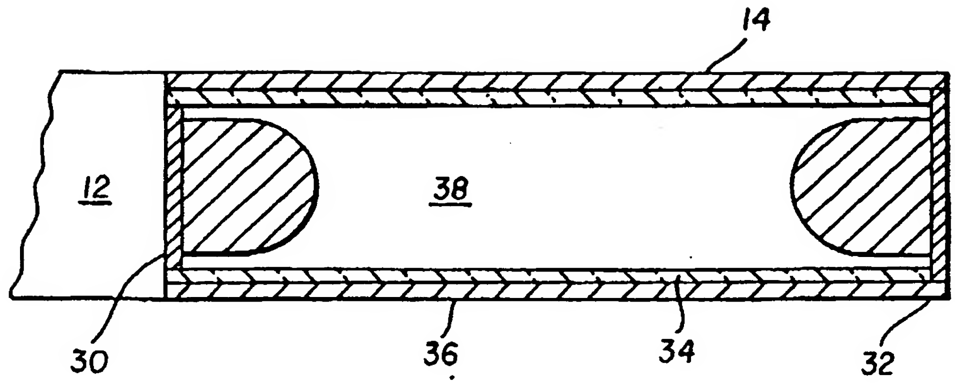


FIG. 2

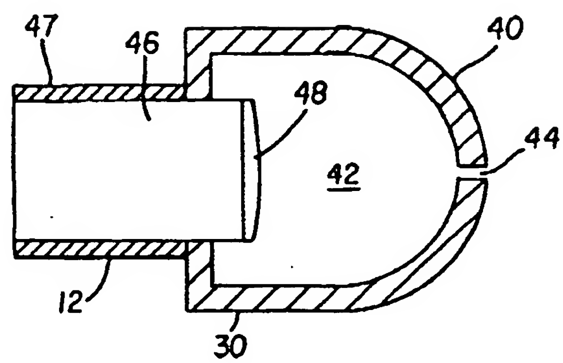


FIG. 3

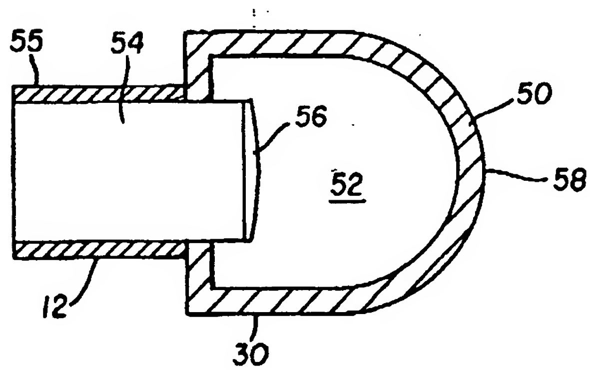


FIG. 4

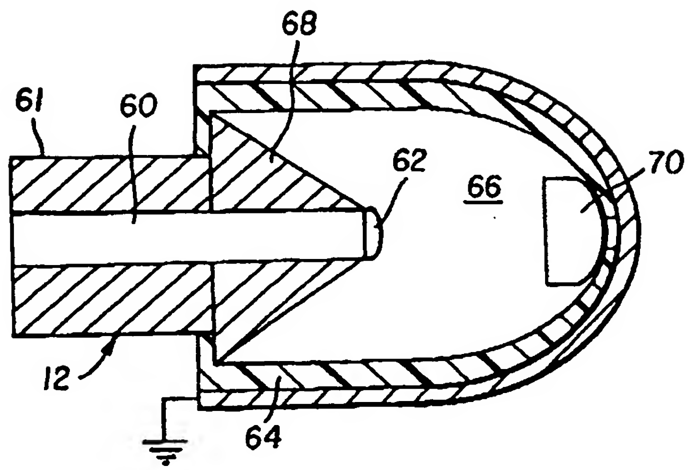


FIG. 5

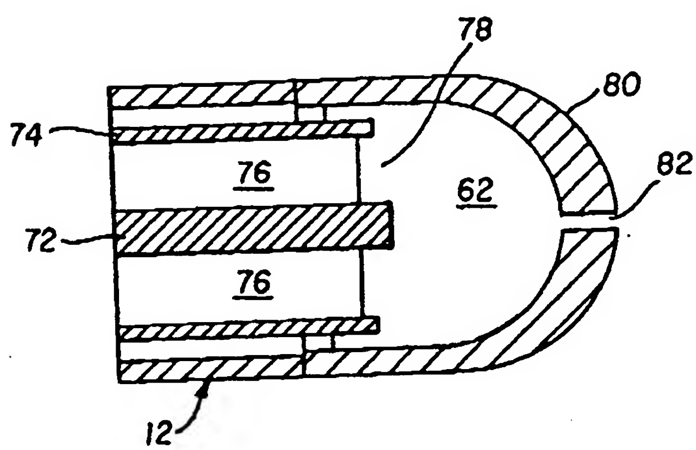


FIG. 6

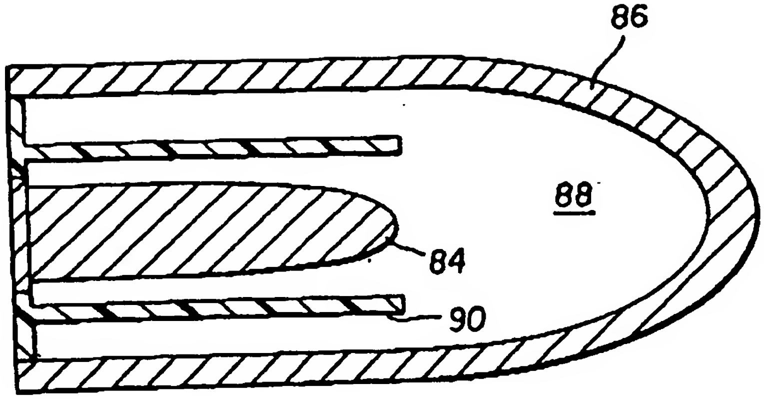


FIG. 7

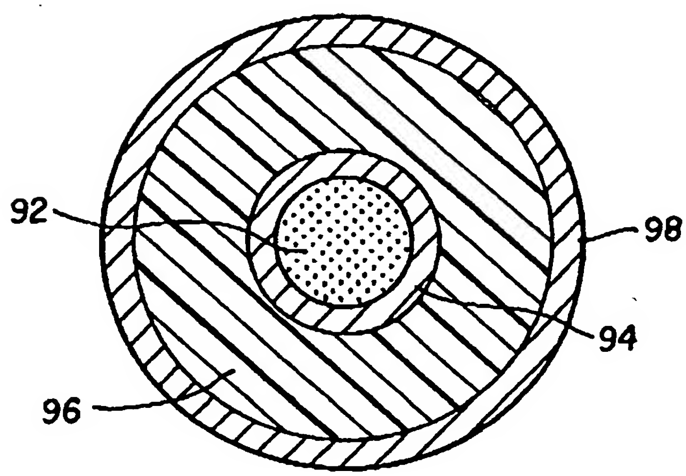


FIG. 8

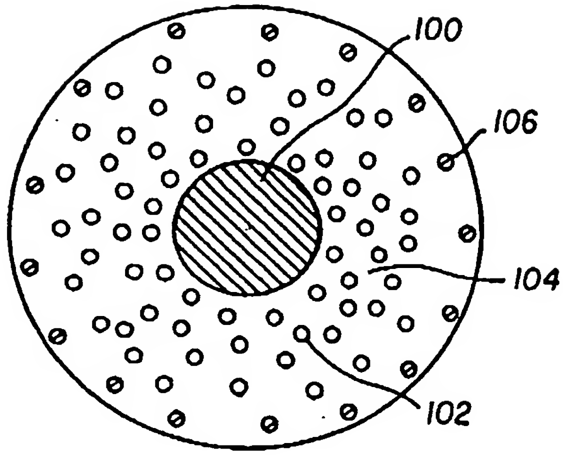


FIG. 9

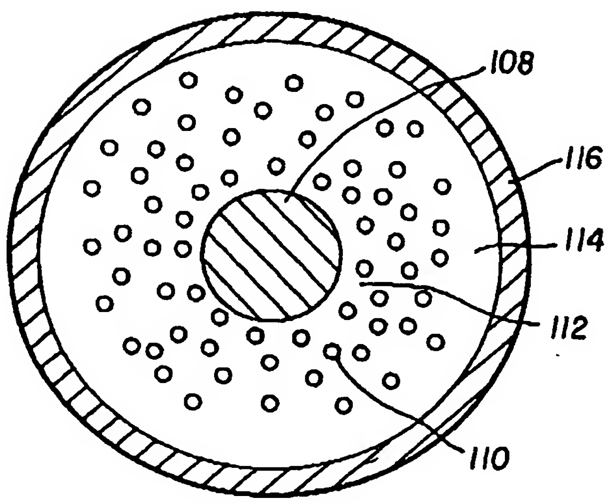


FIG. 10

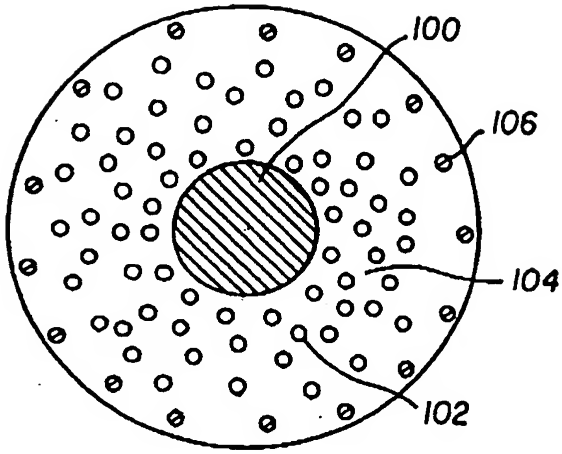


FIG. 9

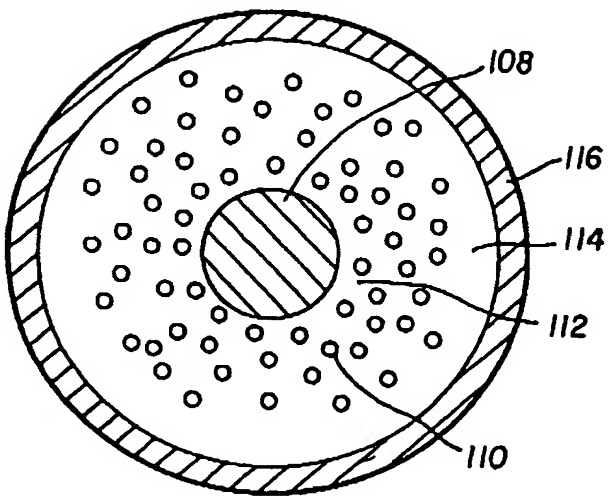


FIG. 10

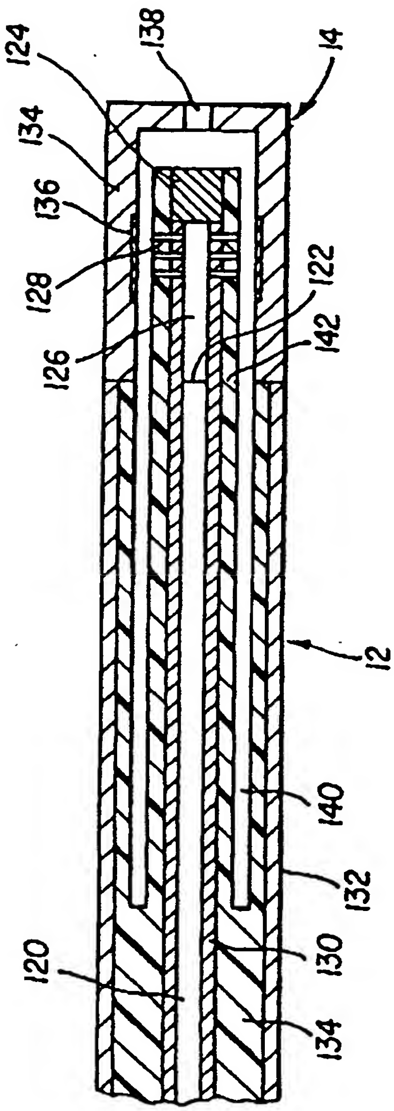

124
138
14
134
136
128
126
122
142
12
140
132
120
130
134

FIG. 11

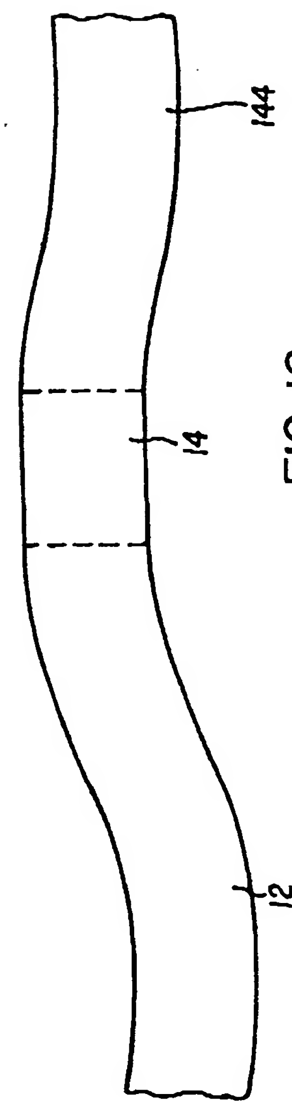

144
14
12

FIG. 12

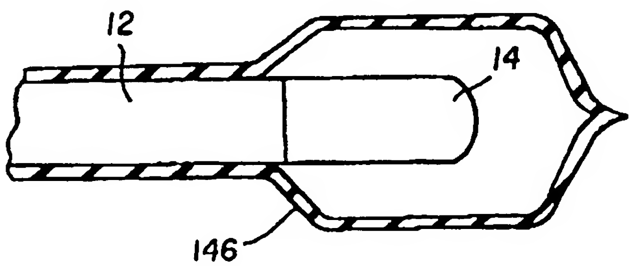


FIG. 13

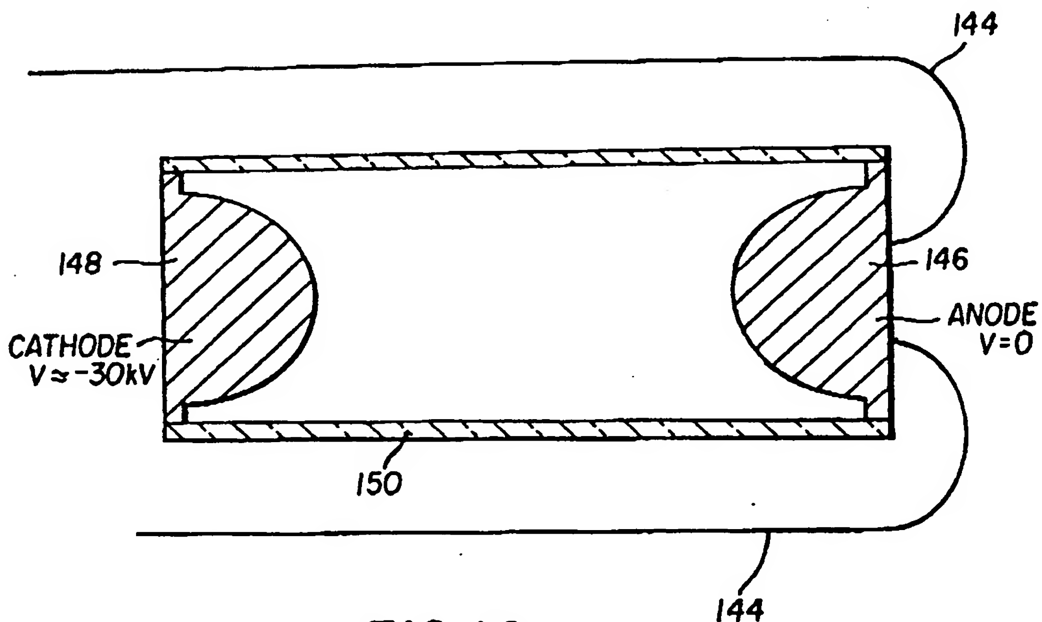


FIG. 14